Investigation of NODDI estimates at two different magnetic fields along the rat corpus callosum

Nicolas Kunz[1], Stéphane Sizonenko[2], Petra Susan Hüppi[2], Rolf Gruetter[1,3], and Yohan van de Looij[4]

[1]*CIBM-AIT, EPFL, Lausanne, Vaud, Switzerland,* [2]*Division of Child Growth and Development, University of Geneva, Geneva, Switzerland,* [3]*Department of Radiology, University of Geneva and Lausanne, Lausanne, Switzerland,* [4]*University of Geneva, Division of Child Growth and Development, Geneva, Switzerland*

TARGET AUDIENCE: Diffusion imaging (DTI and NODDI) at ultra-high magnetic field (9.4 and 14.1T)

PURPOSE: It has been shown that magnetic field strength (B_0) has an influence on diffusion tensor imaging (DTI) derived parameters [1 and references therein], limiting multicentre comparisons with different MR systems. Recently, new diffusion models have been developed such as the neurite orientation dispersion and density imaging (NODDI [2]), a new biophysical diffusion compartment model for estimating the microstructural complexity of neurites (*i.e.* dendrites and axons), successfully used *in vivo* on clinical MRI scanners [2,3]. NODDI provides estimation of microstructural parameters such as intra-neurite volume fraction (f_{icvf}) and orientation dispersion index (ODI) to model the dispersion/fanning of the axonal fibers or dendrites within a voxel. The aim of this work was to investigate effect of B_0 and diffusion time (t_{diff}) on NODDI estimated microstructural parameters to better understand this field dependency.

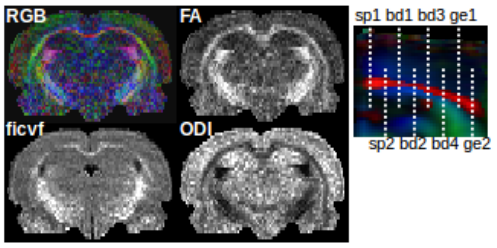

***Fig 1**: DTI color-coded map (RGB), Fractional anisotropy (FA), intra-axonal volume fraction (ficvf) and orientation dispersion index (ODI) of a typical data-set acquired at 9.4T. Right panel: position of the 8 axial slices overlaid on a coronal slice for ROI delineations: splenium (sp1&2), body (bd1,2,3&4) and genu (ge1&2).*

MATERIALS AND METHODS: MR experiments were performed on two MR scanners of 9.4T and 14.1T (Agilent) equipped with high performance gradient coils (400mT/m, 120µs) with a quadrature transceive 20-mm surface RF coil. 4 rats were scanned on each MR system. A multi-b-value shell DWI protocol was acquired using a semi-adiabatic double spin-echo EPI 4-shots sequence [4] with the following parameters: FOV = 23×15 mm^2, matrix size = 128×64, 8 slices of 0.8 mm thickness in the axial plane, 6 averages with TE/TR = 42/2000 ms given an acquisition time of 90 min. A total of 54 DWI were acquired, three of them were b_0 reference images. The remaining 51 were separated in 2 shells with the following distribution (# of directions/*b*-value in s/mm^2): 21/1000 and 30/2000. All 51 directions were non-collinear and uniformly distributed in each shell. Two different diffusion times (t_{diff}) were investigated for each B_0: 13 ms and 20 ms. Acquired data were fitted using conventional diffusion tensor (DT) and NODDI with the NODDI matlab toolbox [2]. ROIs were manually delineated in the rat corpus callosum (CC) at 8 different image planes (*i.e.* one ROI on each slice, Fig 1).

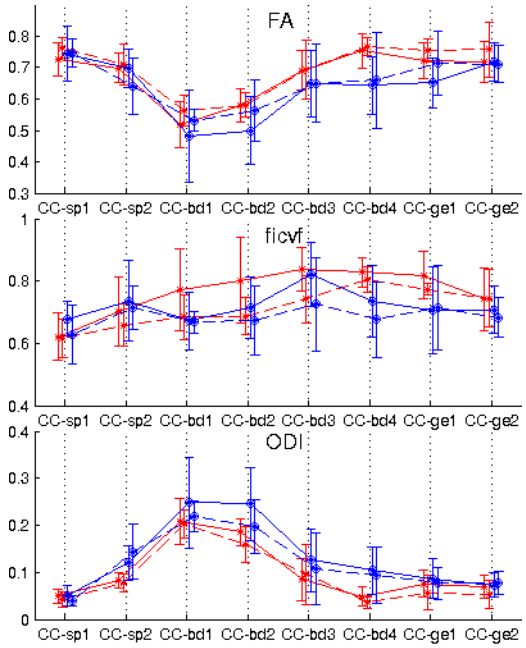

***Fig 1**: Profiles of fractional anisotropy (FA, top), intra-axonal volume fraction (ficvf, middle) and orientation dispersion index (ODI, bottom), along the corpus callosum. Red-solid line: 9.4T t_{diff} =13ms; Red-dashed line: 9.4T t_{diff} =20ms; blue-solid line: 14.1T t_{diff} =13ms; blue-dashed line: 14.1T t_{diff}=20ms. Error bars show the standard deviation over the animals.*

RESULTS AND DISCUSSION: Overall, the diffusion weighted images were of good quality at both magnetic field strengths and diffusion times (Fig 1). The acquisition protocol allowed the reconstruction of DT and NODDI models on the same data-set, with acquisition time suitable for *in-vivo* application.

Corpus Callosum microstructure: In all four conditions, the microstructural changes along the CC were visible in the FA values, with smaller values in slices CC-bd1&2 (Fig 2), where the axons of larger diameter (motor axons) are present [1,5]. Surprisingly, the intra-axonal volume fraction did not show any significant changes along the corpus callosum. ODI, on the other hand, showed the opposite behavior than FA with larger orientation dispersion in the two first slices of the body of the CC (Fig 2, CC-bd1&2). The drop of FA was driven by both a decrease of the axial diffusivity with an increase of the radial diffusivity. The NODDI parameters suggest that these changes arise not only from the axonal diameter but also from a microscopic disorganization in the CC-bd1&2 (increase of the ODI) probably due to an increased heterogeneity in axons dimensions.

B_0 and t_{diff} dependency: Due to the low number of animals per group, no significant differences were visible between the two magnetic field strengths or diffusion times. However similar trends reported in a previous study [1] showing higher FA values at longer t_{diff} and a small reduction of FA at higher B_0 were visible (Fig 2). ODI parameters showed the same kind of dependency but with smaller dispersion at longer t_{diff} and higher dispersion at stronger B_0.

CONCLUSIONS: We demonstrate the feasibility of reconstructing NODDI model in the rodent brain *in-vivo* at ultra-high magnetic field using multi-b-value shells acquisition. These preliminary results suggest that FA changes along the CC are not only due to differences in axonal diameter but also to axonal orientation dispersion differences as depicted by NODDI results. Number of animals per condition will be increased to address the B_0 and t_{diff} dependencies and to confirm the trends observed in this study.

References: [1] Kunz N. NMR in Biomed. 2013; [2] Zhang H. NeuroImage 2012; [3] Kunz N. NeuroImage 2014; [4] van de Looij Y. Magn Res Med 2011; [5] Barazany D. Brain 2009.

Acknowledgements: Supported by the CIBM of the UNIL, UNIGE, HUG, CHUV, EPFL, Leenards and Jeantet foundation.